Analysis of Sealing vs Tensile Bond Strength of Eight Adhesive Restorative Material Systems

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Purpose: Using a simulated perfusion system, the intent was to determine: 1) the sealing ability of eight restorative materials (five composite resins and three compomers) used together with their corresponding dentin bonding systems, 2) their tensile bond strength, and 3) the correlation (if any) between both parameters.

Materials and Methods: Permeability in crown segments from sound human third molars (n = 140) subjected to simulated perfusion (32 cm of distilled water) was measured before and after restoring with each material, and the percentage of decrease in permeability (PPD) was recorded. Specimens were later subjected to tensile tests to determine the tensile bond strength (TBS) of the interface. Finally, parameters were analyzed for correlation.

Results: For the eight adhesive systems, the percentage of reduction in permeability was relatively high [mean in %, (SD)]: F2000 93.6 (5.7), SB1 88.6 (11.3), SSC 86.0 (5.7), PB20 81.1 (15.9), COM 77.5 (10.8), OPTS 75.3 (20.6), DYR 73.7 (12.7), SSPR 65.5 (19.8). TBS values were relatively low [mean (SD)], in MPa: F2000 1.8 (0.7), SB1 4.9 (1.4), SSC 2.6 (1.4), PB20 4.3 (1.2), COM 2.4 (1.1), OPTS 4.5 (1.7), DYR 1.6 (0.6), SSPR 4.2 (1.5). We could not demonstrate any statistically significant correlation between both parameters for these results (maximum significance [F2000]: $r = 0.39$, $p = 0.206$).

Conclusion: No material completely ceased to filtrate through the interface. The low TBS values were probably due to the large size of adhesive areas. No significant correlation was found between PPD and TBS for the materials tested. There was a statistically significant relationship ($r^2 = 0.063$, $p = 0.018$) between TBS and TBA (total bonded area), described by the equation $TBS = 5.9 - 0.03 \cdot TBA$.

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One of the functions of adhesive materials is to reseal the tooth and restore the tooth's integrity damaged by caries, fractures, erosions, or cavity preparations, thus preventing the entry of contaminants and microorganisms from the oral environment and a detrimental level of fluid movement through such interfaces.

Dentin is a mineralized tissue, the entire thickness of which is occupied by dentin tubules at densities varying between 20,000 and 80,000 per mm^2, depending on the tooth and area studied. Tubule diameter ranges from 0.8 to 2.5 µm.[2,38,47,49] These tubules are filled with, among other things, the dentinal fluid, an extracellular fluid[13,28] similar in composition to plasma,[23,46] synovial fluid, or cephalorhachidian fluid.[40] This fluid probably derives from the odontoblastic capillary plexus.[42]

These structural peculiarities of dentin make it a permeable tissue where the movement of fluid between the pulp chamber and the external surface

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can occur, due to the difference of hydrostatic pressure between the two.[23] According to Brännström, the sudden acceleration or deceleration of fluid displacement[1,5] provokes the sensitivity that appears when the protective enamel is lost. The movement of fluid towards the outside also produces a moist surface that could affect the bonding mechanisms of the adhesive materials.[17,24,44]

The dynamics of this fluid movement and its influence on adhesion mechanisms of the restorative materials have been studied by Derkson, Pashley, and Derkson.[8] They described a system of measuring the potential for microleakage of different restorative materials in vitro simply by comparing the permeability of dentin before and after filling. This method cannot perfectly reproduce dentin's physiology, but it does provide an approximation of the real clinical conditions.

To study the sealing capacity of adhesive restorative materials, many other systems have been used over the years. One of the most popular is to immerse the specimen to be analyzed in different staining solutions, and observe its diffusion throughout the restorative material–tooth interface.[9,10,32] The large size of the molecules of some of these dyes requires there be a sufficiently wide gap to allow the diffusion through the interface. There are, however, other studies using silver nitrate which reveal the presence of a porous zone under the interface that – in the absence of a measurable gap – is potentially permeable to oral and/or dentinal fluids.[37] These types of methods apply a numeric scale in terms of the depth of penetration of the dye, meaning it is a qualitative method with results that are difficult to standardize. Also, the degree of staining is influenced by factors such as the concentration and type of colorant, its affinity for dentinal tissues and/or restoration materials, and time of permanence in the solution, among others.[46] Although very sensitive in detecting microfiltration given the small size of their molecules, systems based on penetration of radioactive isotopes show disadvantages similar to the ones already mentioned.[46]

Studies based on SEM are the only ones to date that allow the ultrastructure of the interface to be observed. However, there are some problems associated with the method. The preliminary handling of the specimens (dehydration, fixing, drying, sputtering, etc) could cause alterations in the structures studied leading to images that do not reflect reality. Additionally, the small size of the samples studied does not allow generalization of observations made even within same specimen.[7]

One of the present trends in research on dentin permeability and its application in studies of adhesive materials is based on the work of Pashley and colleagues[8,34,45] and others.[24,48] These authors simulate in vitro the pressure and humidity conditions of the dental tissues. This methodology allows sealing studies of the restorative materials to be performed under humid conditions more closely resembling the oral environment, a decisive factor in forming the interface,[15,29] and makes it possible to quantify and compare results. Using this system, adhesion studies can also be conducted in which the strength of an interface created in a specimen subject to simulated pressure is determined under conditions more like those found in vivo.

There are basically two kinds of fluid used by the different authors to fill the perfusion system: fluids with[14,21,25,43] and without[8,11,24,33,34,44] protein. Tagami et al[43] carried out a study comparing the sealing capacity of different adhesive materials in terms of the perfusion liquid used: phosphate buffered saline or bovine serum. They observed that sealing capacity was significantly higher when the liquid used was bovine serum, attributing the drop in permeability to the precipitation of serum proteins, due to the presence of the resin primers inside the tubules. In this context, reports on conductance performed with immunoglobulin solutions in distilled water demonstrated the potential mechanism of immunoglobulins in decreasing permeability.[14] Nikaido et al[25] obtained better tensile results with bovine serum than saline serum as the perfusion liquid.

The mechanical strength of the adhesive interface has often been studied in vitro, mostly using air-dried dental tissues[3,6,16,31,35,36,39] and yielding results that may not accurately reflect in vivo conditions because they do not consider the presence of fluid in dentin and its possible influence in forming the hybrid layer. Other authors have tried to reproduce in vitro the physiological conditions of pressure and humidity of the tooth, using simulated perfusion systems to carry out studies about the mechanical characteristics of restorative material–tooth bonding.[24,25,33,44]

The potential effect of such fluid on the correlation between the sealing properties of the adhesive interface and adhesion has been studied by various authors, with different results.[34,45] Studies on the shear bond strength of some of the materials (or

similar ones) used in this examination show ranges of mean TBS (in MPa) in non-perfused teeth of 10.9[20] to 30[19] for Single Bond (Scotch Bond 1 in Europe), 12.2[20] to 23.4[19] for Optibond Solo, and 7[30] to 26.4[41] for Prime & Bond 2.1. In perfused teeth,[21] the TBS for Single Bond was 15 MPa, for Optibond Solo 16.3 MPa, and Prime & Bond 2.0 showed 7.6 MPa. Other studies show mean TBS values (MPa) of 4.9 for Syntac[4] and 14.2 for Syntac Single Component.[41] Vargas and Cobb[50] reported shear bond strength values of 6.9 MPa for Dyract, 8.6 for Compoglass, and 13.8 for F2000, applying the adhesives provided with these compomers.

The purpose of the present study is to determine 1) the sealing ability (expressed as conductance decrease) of eight restorative materials (five composite resins and three compomers) used with their corresponding dentin bonding systems, 2) their tensile bond strength, and 3) the correlation (if any) between both parameters.

MATERIALS AND METHODS

We used a simulated perfusion system derived from the one described by Derkson, Pashley, and Derkson,[8] consisting of a 32-cm column of distilled water – to reproduce the intrapulpal pressure[27] – joined by a flexible tube to a graduated micropipette. At the other end of the micropipette, another flexible tube allowed the fluid to flow through the specimen. Using a microsyringe, an air bubble was introduced in the micropipette. Its displacement towards the tooth allowed us to measure the amount of fluid that was lost through the exposed dentin.

One-hundred forty surgically extracted sound human third molars were preserved in a 70% ethanol solution[11] until use, not longer than one month. Although a total number of 140 third molars were used during the course of this study, not all of them were used at the same time. Several samples were discarded from the study at each step for different reasons: cohesive fractures in dentinal tissue and/or restorative material, difficulties in determining permeability decrease in some samples, impossibility of measuring the dimensions of the bonding area, etc. This explains why the total number of specimens tested was not identical in all parts of the study.

The crowns were separated from the roots at the furca level with a diamond turbine handpiece, thus eliminating the floor of the pulp chamber. The contents of the chamber were carefully extracted with cotton pliers; care was taken to avoid touching the pulpal roof. The occlusal enamel was removed with 120-grit sandpaper (Struers, Rødovre, Copenhagen, Denmark) mounted on a polisher (Struers Dap-7, Rødovre, Copenhagen, Denmark) with constant water cooling, exposing a dentin area surrounded by peripheral enamel.

Rectangular polymethyl methacrylate (PMMA) bases 1.5 mm thick were prepared with a central perforation. Each base was affixed with a cyanoacrylate adhesive (Super Glue 3, Loctite, Spain) to the radicular face of the prepared crown. To minimize filtration through noncontrolled surfaces, the lateral area was coated with a layer of nail varnish (Natural Wonder Nail Varnish, Revlon, Barcelona Spain).

Specimens were embedded in resin (Chronolite 10700 + activator 3015, Plastiform, Madrid, Spain) in the lower part of a pair of stainless steel cylinders and flush to it. After embedding the specimen, the area to be studied was polished with 600-grit sandpaper.

The upper part of the cylinders formed a matrix, framing the exposed area to permit a subsequent restoration to be made (Fig 1).

The entire assembled unit (resin-tooth on lower cylinder and resin matrix on upper cylinder) was connected to the perfusion system. The pulp chamber was filled with water and joined to the perfusion system by introducing and sealing a needle through the hole left in the embedding resin and in the PMMA base. Before measuring, specimens were connected to the system for 24 h to allow complete rehydration of dentin.

The permeability of the specimen was then determined, measuring the amount of liquid that passed through it during a minimum period of 30 min.

Subsequently, restorations were made of the materials tested (Table 1) so that the entire exposed tooth surface was subjected to adhesion. Each test group consisted of a restorative material plus associated adhesive, as follows:

- COM (Compoglass plus SCA, Ivoclar/Vivadent, Schaan, Liechtenstein)
- DYR (Dyract plus PSA, Detrey/Dentsply, Konstanz, Germany)
- F2000 (F2000, 3M, St Paul, MN, USA)
- SSC (Tetric plus Syntac SC, Ivoclar/Vivadent)

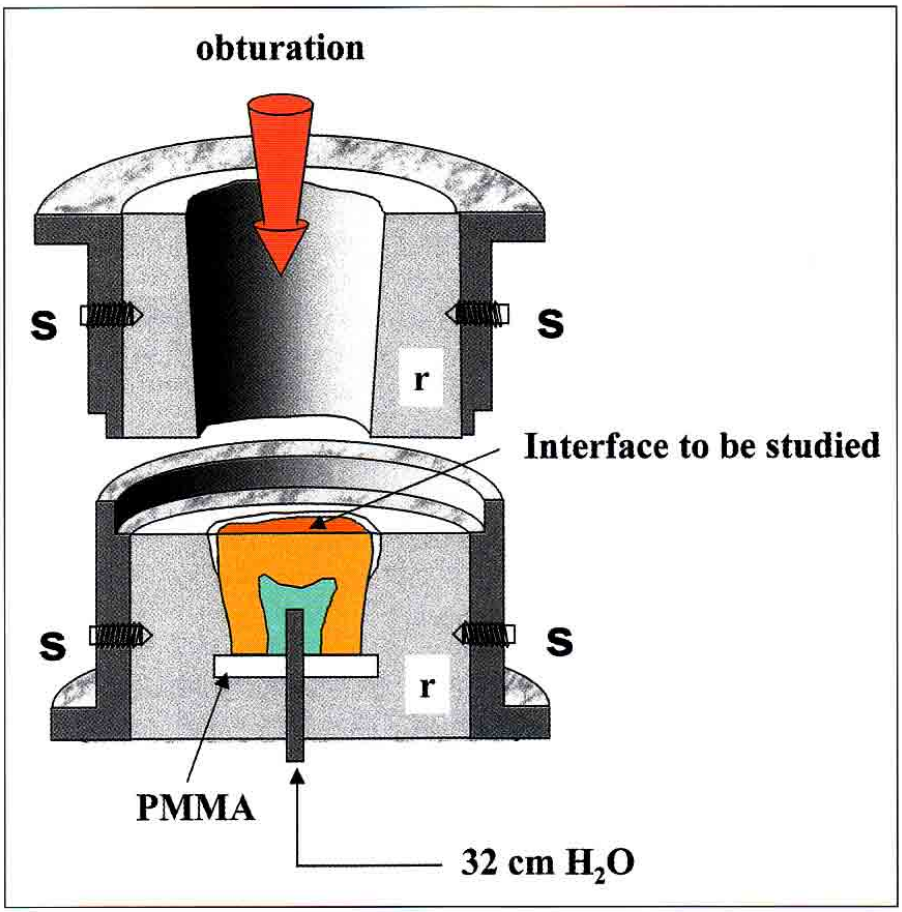

Fig 1 Sagittally centered section of setup. PMMA: polymethyl methacrylate base glued to the specimen. s: screws to fix the embedding resin to the metallic rings. r: embedding resin.

Table 1 Materials and usage

Group	Adh. system	Manufacturer	Etch	Wash/Dry	1st layer	VLC	2nd layer	VLC	Rest. material	Manufacturer	Restoration
COM	SCA	Ivoclar/Vivadent	-	-	8–12-14	16	17–18	16	Compoglass	Ivoclar/Vivadent	19–20
DYR	PSA	Detrey/Dentsply	-	-	8–13-14	15	17–18	15	Dyract	Detrey/Dentsply	19–20
F2000	F2000	3M	-	-	8–13-14	-	17–18	15	F2000	3M	19–20
SSC	Syntac SC	Ivoclar/Vivadent	1	3–6	8–12-14	16	17–18	16	Tetric	Ivoclar/Vivadent	19–20
PB20	Prime & Bond 2.0	Detrey/Dentsply	2	5–6	8–13-4	15	17–18	15	TPH Spectrum	Detrey/Dentsply	19–20
SB1	Scotch Bond 1*	3M	1	4–7	8	-	17–18	15	Z100	3M	19–20
SSPR	Syntac Sprint	Ivoclar/Vivadent	1	3–6	9–11-14	-	17–18	-	Tetric Ceram	Ivoclar/Vivadent	19–20
OPTS	Optibond Solo	Sybron/Kerr	1	3–6	10	-	17	16	Prodigy	Sybron/Kerr	19–20
									TPH Spectrum	Detrey/Dentsply	19–20

KEY: *Equivalent to Single-Bond sold in USA. 1: Total etch 37% Orthophosphoric acid (Email Preparator GS; Ivoclar/Vivadent) 15 s; 2: Total etch 37% Orthophosphoric acid (Email Preparator GS; Ivoclar/Vivadent) 20 s; 3: Extensive wash; 4: 10 s wash; 5: 15 s wash; 6: gently air dry; 7: 1 to 2 s air dry; 8: Apply adhesive; 9: 10 s brush adhesive; 10: 15 s brush adhesive; 11: 15 s wait; 12: 20 s wait; 13: 30 s wait; 14: gently air dry; 15: 10 s VLC; 16: 20 s VLC; 17: 2nd layer; 18: immediate gentle air drying; 19: 1st layer (<2mm) and 40 s VLC; 20: 2 or 3 layers (< 2 mm) and 40 s VLC per layer.

- PB20 (TPH Spectrum plus Prime & Bond 2.0, Detrey/Dentsply)
- SB1 (Z100 plus Scotch Bond 1, 3M)
- SSPR (Tetric Ceram plus Syntac Sprint, Ivoclar/Vivadent)
- OPTS (Prodigy or TPH Spectrum plus Optibond Solo)

When Optibond Solo (Sybron/Kerr, Orange, CA, USA) was tested, 15 specimens were restored with Prodigy (Sybron/Kerr, Orange, CA, USA) and 7 with TPH Spectrum (Detrey/Dentsply, Konstanz, Germany). This was due to the large number of cohesive fractures of Prodigy when tensile tests were performed.

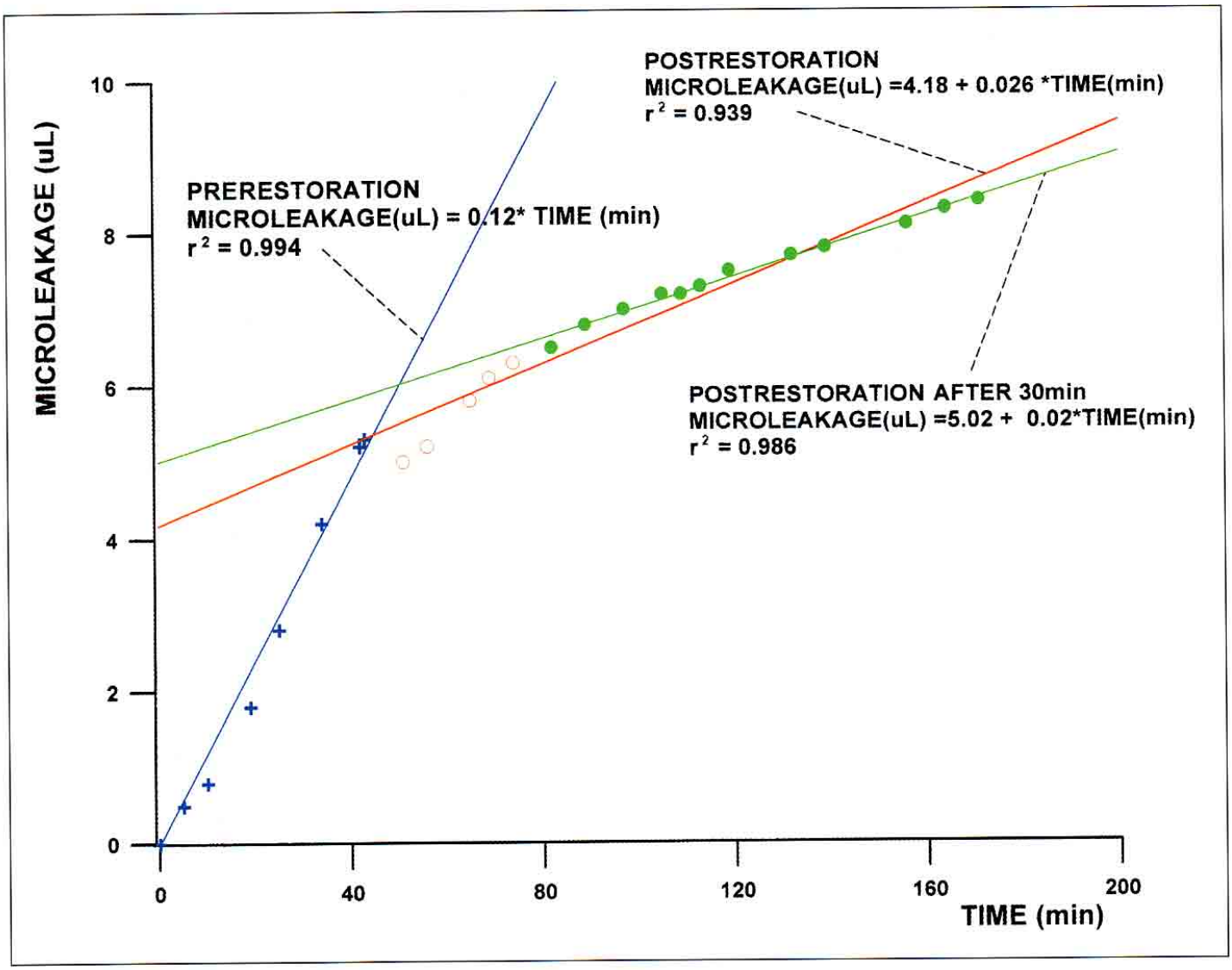

Fig 2 Flow curve of a specimen. Prerestoration: regression line of flow of specimen before bonding procedure (+). Postrestoration: regression line of whole flow line after bonding (empty and filled circles). Postrestoration after 30 min: regression line of flow after bonding excluding data of the first 30 min (empty circles).

Postrestoration permeability was determined in all specimens over a period ranging from 60 to 120 minutes.

Permeability was measured and recorded as volume per unit of time (flow). Each specimen was described by the slope of the flow line. The first, prerestoration part of the recording (PRE) was assigned a value of 100% permeability.

For each specimen, the percentage of permeability decrease (PPD) postrestoration (POST) was calculated as the percent decrease in the slope of its recording line with respect to the prerestoration measurement (PRE), ie, PPD = 100 x (PRE – POST)/PRE.

For materials where etching was performed as a part of the bonding procedure (SSC, PB20, SB1, SSPR, OPTS), the first 30-min section of the postrestoration line was ignored, considering it would correspond to the rehydration phase of the tooth after rinsing and drying steps, and did not follow the linear pattern of the rest of the measurements (Fig 2).

Immediately after concluding permeability measurements, the assembled cylinders were disconnected from the perfusion system, placed in a mechanical testing machine (H 5000M, Hounsfield, Croydon, UK), and subjected to traction (1 mm/min) perpendicular to the interface using a 500-N cell until they debonded. All specimens exhibiting cohesive fractures in the filling material and/or dental tissues were excluded from the study.

The adhesive areas were measured with an

Table 2 Total bonded area (mm^2)

Adhesive system	n	TBA (SD)
F2000	12	85.68 (11.52)
SB1	16	85.53 (12.08)
SSC	11	88.22 (14.17)
PB20	13	87.41 (15.83)
COM	14	84.26 (10.88)
OPTS	10	78.12 (18.16)
DYR	12	89.96 (9.34)
SSPR	9	77.52 (11.55)

Table 3 Conductance values for teeth ($\mu L/min \cdot mm^2$)

	n	Conductance (SD)
Prerestoration	101	0.00245 (0.00106)
Postrestoration *	38	0.00047 (0.00024)
Postrestoration	57	0.00037 (0.00027)

* Specimens did not require acid etching prior to the application of the adhesive system.

image analyzer (Leica Qwin Q500 IW, Cambridge, UK), obtaining the total bonded area of each sample (TBA) in mm^2. From these area measurements and the results obtained in the tensile test, the tensile bond strength (TBS) of the interface was calculated in MPa.

ANOVA and Newman-Keuls tests were used for the statistical analysis.

For each material tested, linear regression between PPD and TBS was performed to determine whether these variables were correlated.

In order to establish the influence of PPD and TBA on TBS, multiple regression analysis was performed (SPSS 9.0.1, Chicago, USA), in which the dependent variable was TBS and the independent variables were TBA and PPD. This analysis measures the individual influence of each independent variable excluding the bias introduced by the other. PPD was selected as one probable predictive parameter of TBS because it seems reasonable that a permeability decrease induced by an adhesive system is a reflection of the tightness of the bond produced. TBA is known to influence TBS.[36]

RESULTS

Values of total bonded area (TBA, in mm^2) are listed in Table 2.

Pre- and postrestoration values for conductance (in $\mu L/min \cdot mm^2$) of specimens are listed in Table 3. Figure 3 contains boxplots of conductance before and after restoration (in etched and nonetched specimens).

To analyze the change in conductance, paired t-tests were carried out for all materials. All materials exhibited a statistically significant change ($p < 0.01$).

Table 4 presents PPD values for each material. The ANOVA showed statistically significant differences between the groups ($p < 0.05$). The Newman-Keuls test determined these differences ($p < 0.001$) accordingly: F2000 > DYR, F2000 > SSPR, SB1 > SSPR, and SSC > SSPR.

The TBS (MPa) data for all materials are given in Table 5. The ANOVA showed there were statistically significant differences ($p < 0.001$) between the groups. The Newman-Keuls test ($p < 0.001$) identified the following differences: SB1 > COM, SB1 > DYR, SB1 > F2000, and SB1 > SSC; OPTS > COM, OPTS > DYR, and OPTS > F2000; PB20 > DYR and PB20 > F2000; SSPR>DYR and SSPR > F2000. Figure 4 shows boxplots of TBS for all materials.

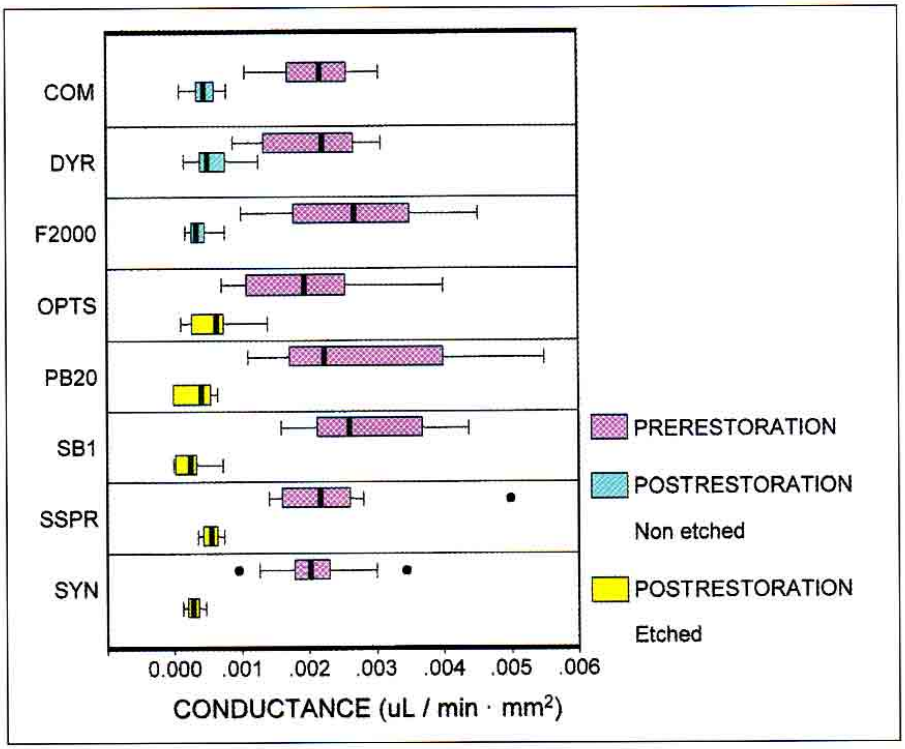

Fig 3 Conductance before and after restoration (in etched and nonetched specimens).

Table 4 PPD for bonded restorations

Adhesive system	n	PPD (SD)	Group*
F2000	14	93.6 (5.7)	A
SB1	18	88.6 (11.3)	A,B
SSC	14	86.0 (5.7)	A,B
PB20	24	81.1 (15.9)	A,B,C
COM	17	77.5 (10.8)	A,B,C
OPTS	22	75.3 (20.6)	A,B,C
DYR	16	73.7 (12.7)	B,C
SSPR	15	65.5 (19.8)	C

* The materials with the same capital letter are homogeneous groups according to the Newman-Keuls test ($p < 0.001$): F2000 > DYR, F2000 > SSPR, SB1 > SSPR and SSC > SSPR.

Table 5 Tensile bond strength (MPa) of adhesives bonded to dentin

Adhesive system	n	TBS(SD)	Group*
SB1	16	4.9 (1.4)	A
OPTS	10	4.5 (1.7)	A,B
PB20	13	4.3 (1.2)	A,B,C
SSPR	9	4.2 (1.5)	A,B,C
SSC	11	2.6 (1.4)	B,C,D
COM	14	2.4 (1.1)	C,D
F2000	12	1.8 (0.7)	D
DYR	12	1.6 (0.6)	D

*The materials with the same capital letter are homogeneous groups according to the Newman-Keuls test ($p < 0.001$) test: SB1 > COM, SB1 > DYR, SB1 > F2000, and SB1 > SSC; OPTS > COM, OPTS > DYR, and OPTS > F2000; PB20 > DYR and PB20 > F2000; SSPR > DYR and SSPR > F2000.

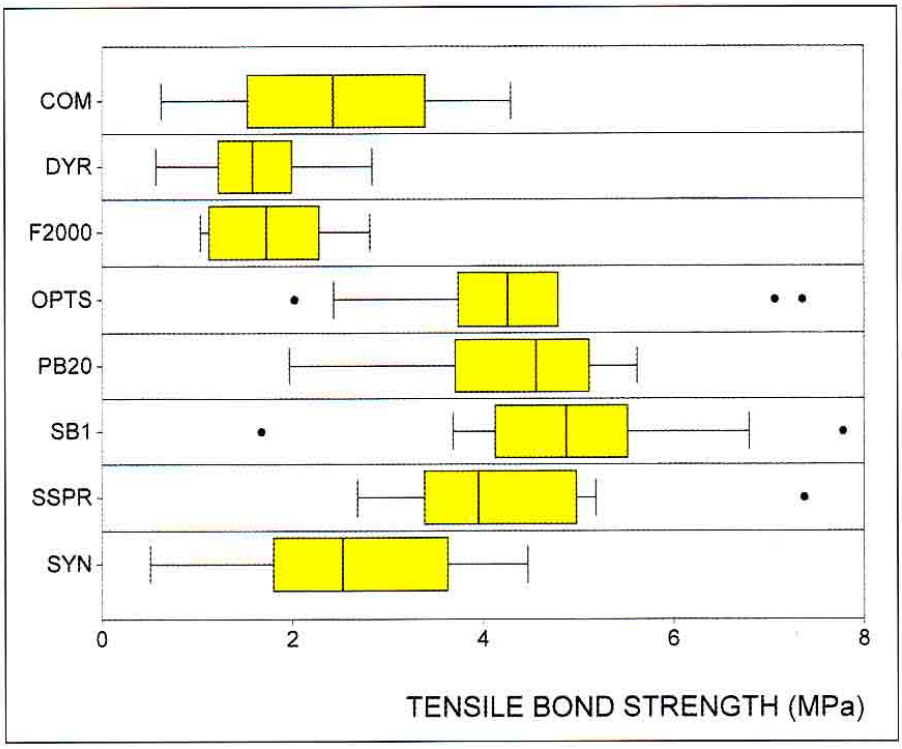

Fig 4 Tensile bond strengths. Values in MPa.

Table 6 Correlation PPD/TBS

Adhesive system	r	p
DYR	-0.33	0.288
COM	0.11	0.690
F2000	0.39	0.206
SSC	-0.25	0.506
PB20	0.29	0.328
SB1	0.09	0.757
SSPR	-0.20	0.629
OPTS	-0.40	0.317

r: Pearson's linear correlation coefficient. p: level of significance

For all materials, the TBS and PPD were analyzed for correlation (Table 6).

The results of the multiple regression analysis are shown in Table 7.

DISCUSSION

We found statistically significant reductions ($p < 0.01$) in conductance by sealed compared to unsealed samples, but no material tested completely stopped filtration. Authors of similar studies also found residual, postrestoration permeability.[34,45,48] The clinical significance of this statistically significant decrease in conductance is not clear. It has yet to be determined what PPD is desirable or even possible.

Residual postrestoration permeability could be due to the presence of hydrophilic resins in these materials. A study performed by Yap and Lee[51] showed that the proportion of HEMA in the adhesive material has a significant influence on the amount of water it can absorb.

The lateral areas of the specimens in this study

Table 7 Multiple regression analysis results

		ANOVA		Regression results				
							95% confidence interval of estimated parameters	
Model	r^2	F	Significance (p=)	Parameters	Estimated coefficients	Significance (p=)	Inf	Sup
1	0.063	5.811	0.018	Constant	5.9	< 0.0001	3.7	8.1
				TBA	0.03	0.018	-0.06	-0.006
2	0.075	3.487	0.035	Constant	4.9	0.001	2.1	7.8
				TBA	-0.03	0.015	-0.06	-0.006
				PPD	0.01	0.286	-0.01	0.04

were noncontrolled filtration sites. Although these areas were covered with a varnish, some degree of permeability may well have been present.

Polymerization shrinkage of the adhesive could lead to the formation of a gap, which may later close due to absorption of water by the material.[18] This can only partially explain the incomplete postrestoration cessation of filtration, because the amount of fluid lost by the simulated system is far greater than the amount of water absorbed by the material tested in the study mentioned above.

The mean initial conductance of our specimens was 0.245 $\mu L/min \cdot cm^2$, using a pressure of 32 cm of distilled water. Reported values in comparable experiments are sparse, and range (in the same units) from 0.0177625 to 0.0086521 with zero pressure. Using a pressure of 350 cm H_2O, Tao[45] reported a conductance of 12.2 $\mu L/cm^2 \cdot min$ per cm H_2O, from which a conductance of 1.11 $\mu L/cm^2 \cdot min$ can be calculated (supposing a constant relation) for a pressure of 32 cm H_2O. Storing our specimens in 70% ethanol could have altered the perfusion rate of dentin due to a possible dissolution of lipid membranes within the tubules.

As mentioned above, the perfusion liquid used in this study was distilled water with no protein content; as indicated by previous studies,[14,25,43] this factor may help to explain the persistence of flow after restorative materials had been applied.

The TBS values for all materials used in this study were relatively low compared to those obtained by other authors. The mean adhesion area of our specimens was relatively large, ranging from 58.5 to 114.5 mm^2 [mean (SD) = 85.1 (13.1) mm^2], where those of other studies were far smaller, ca 1 mm^2, with tensile values of up to 40 MPa.[3,39] Phrukkanon, using adhesive areas of 3.1 and 1.1 mm^2, found that TBS was significantly higher in the second group.[31] Sano et al had previously established this in a study where the adhesive strengths of different materials were compared in terms of the areas.[36] We agree with them that during tensile tests, the beginning of interface detachment can be facilitated by the presence of defects and/or stress lines in it. These defects are more frequently found in large areas rather than in small ones, as Griffith[12] stated in 1920. Some of our specimens were rejected from the study because cohesive fractures of material and/or tooth structure occurred at an apparent TBS of less than 5 MPa.

In addition, earlier studies with nonperfused specimens consequently applied the adhesive system to air-dried dentinal tissues. Although the materials used here were designed to work on moist areas, it is possible, as several authors have shown, that excess water in the adhesion area could interfere with the polymerization of the resin, negatively influencing the mechanical properties of the interface.[15,29] In our experiment, the amount of water in the adhesion surface was controlled only by following the manufacturers' instructions (Table 1).

The studies mentioned above examined sealing on dentin only, where the adhesion areas of our specimens included dentin with peripheral enamel. However, we consider it unlikely that this noticeably effected tensile strength results. Any effect should have been to increase the tensile strength, which, as shown by our low TBS values, was not the case.

Our results revealed no statistically significant correlation between PPD and TBS for the materials tested. Tao and Pashley,[45] in a study performed with Scotchbond and Silux (3M, St Paul, MN, USA)

found a significant correlation coefficient. However, their methods differed from ours: their adhesive areas were smaller, and shearing tests were employed to examine the strength of the interface. These differences make a direct comparison difficult. Although Prati et al[34] were also able to establish this correlation using various materials, their results are likewise not directly comparable to ours, since they disconnected the specimens from the simulated pressure system before restoration.

Multiple regression analysis was performed to determine whether a relationship existed between the dependent variable TBS and the independent variables TBA and PPD, pooling all materials. The two analysis models evaluate the contribution of each independent variable (TBA and PPD), and their results can be interpreted as the effect of each independent variable controlling for the effect of the other (Table 7).

The first model (r^2 = 0.063, p = 0.018) is TBS = 5.9 – 0.03 · TBA. Thus, TBS can be calculated when the TBA is known (in this study, 58.5 to 114.5 mm^2). This seems reasonable, given the formula by Sano et al[36] of TBS = 58.758 – 27.9·$\log_{10}$(TBA) which yields predicted TBS values of 9.5 to 1.3 MPa for a 58.5- to 114.5-mm^2 TBA range. It must be borne in mind that the TBA range in our study differs considerably from that in Sano's (0.25 to 9 mm^2).

In the second model, PPD was the predictive variable. This rendered a similar result (r^2 = 0.075, p = 0.035) but with the added information that, with a constant PPD, TBA had an estimated coefficient similar to the first model. Significance of the estimated coefficient of TBA in model 1 and 2 was p < 0.018 and p < 0.015, resp.

The second model shows that TBS is not influenced by PPD (p = 0.286) but by TBA in an inverse manner: the greater the TBA, the lower the TBS by a factor of 0.03. This is also reflected in Table 6: we found no statistically significant relationship between TBS and PPD for any material tested.

CONCLUSION

Although all materials reduced conductance significantly (p < 0.01), none completely stopped (PPD = 100%) fluid flow through specimens.

Tensile bond strengths were relatively low for all materials tested. Although statistically significant differences were found between materials, the clinical relevance of these differences is not clear.

No statistically significant correlation was found between percentage of permeability decrease and tensile bond strength.

There is a statistically significant relationship (r^2 = 0.063, p = 0.018) between TBS and TBA according to the equation TBS = 5.9 – 0.03·TBA.

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